INFLUENCE OF HOSPITAL BED HEIGHT ON KINEMATIC PARAMETERS ASSOCIATED WITH PATIENT FALLS DURING EGRESS

by

Marissa Christman

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STATEMENT OF THESIS APPROVAL

The thesis of **Marissa Christman**

has been approved by the following supervisory committee members:

Andrew S. Merryweather	, Chair	**03/18/15** Date Approved
James C. Martin	, Member	**03/20/15** Date Approved
Donald S. Bloswick	, Member	**03/18/15** Date Approved

and by **Timothy A. Ameel** , Chair/Dean of

the Department/College/School of **Mechanical Engineering**

and by David B. Kieda, Dean of The Graduate School.

ABSTRACT

Understanding the quantitative aspects of kinematic and temporal parameters in fall-prone populations in natural environments is important, particularly in settings replicating hospital environments where patients are often impaired and less familiarized with the layout. Studies indicate fall rates are much higher in these settings than in comparable community settings. The aim of this study was to determine how bed height and side rail presence/type influence fall risk when patients get out of bed unassisted.

Seventy-nine older adults with mobility impairments performed an unconstrained sit-to-walk movement at three randomized bed heights representing low, medium, and high bed conditions. Three side rail conditions were also studied. Temporal and kinematic parameters were obtained from key sit-to-walk movement events using 3D motion capture and ground reaction forces.

There was no evidence that the presence of side rails influenced kinematics. Temporal parameters proved to be most affected by bed heights, particularly in the low bed condition. Velocity and momentum parameters were less significantly affected between conditions. Participants appeared to use similar momentum strategies to rise and initiate gait but altered their timing in order to accommodate

their balance deficits. This study supports the model that suggests increased impairment leads to slower movement event timing during sit-to-walk transition. This study did not support other findings that mediolateral kinematics were higher in those with greater impairments, nor did bed height alter any of these kinematics at any event. Participants had statistically significant higher forward velocities when initiating gait from the medium bed condition, and they had statistically significant lower posterior momenta when exiting the high bed condition. These could be indications of increased mobility and improved use of generated kinetic energy. These represent potentially favorable results in light of reducing fall risk.

Medium bed height appeared to produce the least significant differences in parameters when compared to the two other bed heights. This implies the most flexibility to prioritize postural stability *or* postural mobility. Low bed heights generated particular problems by reducing fluid motion and creating more impediments to postural stability. This suggests that low bed heights may not reduce fall rates during bed exit.

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INTRODUCTION

1.1 Fall Epidemiology

Research centered on fall-prone populations and patient characteristics related to fall risks has produced hundreds of studies and documents but unclear results about a clear directive for a large societal problem. As the human population continues to expand and modern medicine produces longer life spans, the numbers of those at risk for injurious falls due to age and disease increases. While falls can occur at any age, those over 65 have greater rates and more severe consequences, including death. Falls are considered the sixth leading cause of mortality in this group [1-3]. Additionally, injuries sustained from falls in this age group predispose individuals to declines in motor abilities required for day-to-day life, including loss of function in independent living activities and further disease and co-morbidities [4, 5].

Aside from personal suffering, injurious falls create a drain on economic resources, typically requiring, at minimum, short term medical treatment but often acting as the catalyst for long term medical complications. Impairment resulting from falls may create a cycle of recurring falls; regardless of cause, approximately half of all individuals older than 65 suffer such recurring falls [3]. Hamacher et al. [1] concluded in their review that treatment for falls within the U.S., UK, EU, and Australia in the year 2009 cost 0.85-1.5% of total healthcare expenses.

1.1.1 Falls in Hospital

A large body of the literature on falling has primarily examined community-dwelling populations and secondarily examined nursing home populations. Less is known about patient falls in hospitals [6, 7]. However, research has been branching out to this subpopulation and the associated environment. Further underlining the need for this specialized attention is the fact that falls comprise the largest single category of reported incidents in hospitals [8]. Many of these studies have concluded that falls in hospital settings are distinctly different from those in the community. One particular metric that stands out is the rate of falls: randomized studies of older adult community settings have estimated rates of 5 or less falls per 1000 person-days, while similar hospital studies have estimated rates as high as 20 falls per 1000 patient-days in some cases (the latter figure being the high end of a wide discrepancy in reported numbers) [9]. Other factors which set them apart: a hospitalized individual is operating in an unfamiliar environment on a schedule not of their making while under the influence of prescription drugs for other medical problems [10].

Fall rates in hospitals also vary across nursing unit type. Bouldin et al. [11] found that one in four patients who fell in a hospital sustained an injury. However, this rate is not consistent across units; they found patients in medical nursing units had the highest rates of falls with the most severe outcomes. Hitcho et al. [12] also found both medical units as well as neurology units had the highest fall rates within hospitals. Interestingly, these two units also tend to have the highest nurse-to-patient ratios. These findings suggest that nurse-to-patient ratio is neither contributing to nor preventing falls directly. Often these units contain patients who have complicated medical diagnoses but remain mobile. This is a key characteristic associated with the rates of falls in these areas of a hospital.

What happens when in-patients fall? They typically have longer hospital stays and higher medical charges than comparable nonfallers: they appear to remain hospitalized 71% longer and pay 61% higher costs even after controlling for confounders [13]. Sometimes lawsuits may be brought against hospitals by patient families as well. Generally speaking, in-patient falls are an adverse event for all involved. In fact, in autumn 2008, Congress implemented a policy deeming postadmission, in-hospital falls resulting in injury as nonreimbursable under Medicaid and Medicare, making the case that such associated health care costs are part of “Never Events,” i.e., events which should not occur and which are considered preventable [11]. Ironically, despite this legislation, systematic review of the literature on hospital falls by Oliver et al. [14] just a few years prior “has found no consistent evidence for single or multiple interventions to prevent falls.”

1.1.1.1 Etiology

This leads us to consider the characteristics of in-patient falls. Many studies have attempted to holistically examine variables which may contribute or are believed to create fall risks in hospitals. Unfortunately, many hospitals do not record fall rates since only a handful of states have mandatory reporting; therefore, the perceived causes or other elements which may be present at the time of the fall (such as medication usage, patient mental state, clothing or footwear worn, environmental hazards, time of day, people present, tasks being carried out, etc.) and which may influence fall risk factors are often never identified. Likewise, little is known about what might prevent these falls or how effective fall prevention methods already in place actually are, despite the fact that the Centers for Medicare and Medicaid Service have deemed them preventable [12]. Many studies agree that falls are the result of multifactorial risks which involve both intrinsic and extrinsic factors acting together and which are difficult to label discretely as “avoidable” or “not avoidable.” In particular, it is not known if hospital patients at risk for falling demonstrate different manifestations and prevalence of previously-studied risk factors in community populations [7, 9, 12,14, 15].

1.1.1.1.1 Risk Factors and Characteristics

Oliver et al. [14] found a few key characteristics emerge as significantly more prevalent in fallers vs. nonfallers during their analysis of published hospital in-patient fall risk assessments. They were:

- Gait instability
- Lower limb weakness
- Urinary incontinence/frequency or need for assisted toileting
- Previous fall history
- Agitation/confusion or impaired judgment
- Certain prescription drugs, particularly centrally-acting sedative hypnotics

These factors are consistently found despite a nonhomogenous distribution of hospital settings, patient types, and risk factors. Unfortunately, environmental risk factors and patient clinical assessments were not typically examined in the review articles analyzed, so these data appear to be lacking.

Hitcho et al. [16] attempted to collect and characterize fall circumstances over a one-year period at the 1300-bed, academic teaching Barnes-Jewish Hospital in St Louis, MO (excluding physical therapy and psychiatry departments). Among their significant findings were that most falls occurred in the patient's room (85%) while unassisted (79%), citing lost balance as the most common reason (12%). Half of the falls were elimination-related, particularly if the patient was over 65 years old (83% vs. 48% for those younger than 65) although only 19% of these falls actually occurred in the patient's bathroom. The top three activities cited at the time of the fall were ambulation, getting out of bed, and sitting down or standing up. Forty-two percent of falls resulted in some type of injury. The majority of falls involved the patient moving without any staff or device assistance, despite the fact that one-fourth of patients used ambulatory aids at home. Interestingly, injury rates from falls were not statistically different between those over 65 and those under 65 years of age.

Other studies have noted that 50-70% of hospital falls occur in the patient's room, on or near the bed, and are often transfer-related, typically with no assistance or witnesses [5, 16-19]. Using a predictive statistical model of falls supported by routinely available data over a one-year period in a high volume hospital, Halfon et al. [9] reported falls involving the bed ranked as the second most common cause for both first falls and all falls, coming only after falls due to slipping, tripping, and stumbling.

1.1.1.1.2 Prevention

While it is unlikely that all falls can be avoided, it is both possible and necessary to continue efforts at finding simple methods to reduce falls in the most vulnerable populations across a wide variety of in-patient settings. Most fall assessment tools and prevention efforts have focused on modifying individual patient characteristics, such as mobility techniques and exercise-based rehabilitation. When looking at extrinsic factors, they tend to focus on equipment usage instruction and medication adjustments. Unfortunately, many fall risk assessment tools are not designed well in terms of scoring and implementation clarity and consistency in a clinical setting; additionally, many have not gone through any form of validation. Often, these tools are based on risk factors derived from residential-based populations which, as previously noted, may have limited applicability within a sick hospital population. There are some exceptions which demonstrate a streamlined approach and have demonstrated some level of clinical sensitivity and specificity. These include the Morse Fall Scale (MFS) and a few other various fall index scoring methods such as STRATIFY [20, 21]. However, many of these fall index scales require retrieval of data which may not be available in medical records. But the question remains: even if we can properly identify potential problems, can we do anything to change it? So far, analysis of prevention programs has not demonstrated they have any consistent result in reducing falls in hospitals [14, 19].

While research has not completely ignored environmental factors contributing to falls, it appears that little effort has been concentrated on investigating this topic, particularly with regard to the room design. The room layout and equipment are generally the most immediate, recurrent potential hazard to patients [8]. More pointedly, there are few items in a hospital room as omnipresent across clinical spectrums as the bed. With regard to the previously cited statistical incidence of falls within a patient's room and proximal to the bed, it makes sense to explore the interaction between patient and bed. Specifically, bed height and side rail presence/absence has been considered a method of influencing patient fall safety for some time. Despite a lack of empirical support, many nurses and clinicians have felt that very low beds reduce injuries during patient falls – supposedly by limiting how far a patient travels before contacting the floor. On the other end of the spectrum, high bed settings have been used by nurses as a type of restraint imposed on patients to reduce falls; essentially the height will induce reluctance in patients to attempt exiting the bed without staff assistance [22]. This "high bed" practice has been discouraged in many, if not most, hospitals in developed countries as the use of restraint practices may carry its own liabilities, and so maintaining bed heights as low as possible when staff are not present has become the new norm in some clinical settings. But in at least two recent studies, it was concluded that there was no evidence from randomized trials that low-low beds decreased fall rates in any significant way [10, 19, 23].

Likewise, the presence of elevated side rails was often used as a method of bed restraint. Conversely, many healthcare professionals believed that having at least some of the side rails present and elevated actually assisted patients in getting in and out of bed, thereby providing some buffer against the risk of falling by aiding in the vertical transfer movement (this is obviously dependent on the type of side rail used). However, some investigations have shown that bed rail use does not necessarily appear to reduce fall rates and can present a hazard in the form of patient entrapment [5, 16, 24, 25].

Since many falls actually happen during ambulation, it may not seem surprising that low bed heights or the presence of elevated side rails have little impact on fall rates. In such cases, patients have already exited the bed safely and their fall events are likely precipitated by other factors. But the fact that bedside falls still account for a large proportion of reported incidents is a problem. Most of the (very limited) literature directly examining hospital bed heights and fall risks has been published by Tzeng et al. [8, 22, 24, 26] and is qualitative in nature. Yet many qualitative measures lack precision in revealing empirical trends when attempting to determine ergonomic effects on human balance metrics. This view was stated by Salgado et al. [27] when examining predictors of falling in older hospital patients, making the case for more quantitative testing in these types of populations. They found that poor performance during turning on the Timed Up and Go (TUG) was not significantly different between fallers and nonfallers, thus providing little evidence of fall risk with this commonly-used qualitative measure.

An important element of fall research will be to separate out factors which may correlate with or predict falls but which do not necessarily cause them. For instance, fall prevention programs which address predictors only are unlikely to have success in reducing falls. It is possible this may explain why some fall prevention programs do not appear successful in pooled analyses [11]. Further, in-patient fall prevention efforts will undoubtedly be quite different from those in community or long term care facilities since methods such as exercise and other rehabilitation programs are of little use in hospitals where short term patient stays are the norm.

1.2 Biomechanical Analysis

To begin the process of exploring unknowns in the fall literature regarding patient-bed ergonomics, we sought to examine the influence of various bed heights during the sit-to-walk movement in a fall-prone population using quantitative methods. To our knowledge, no study to date has done this. Most quantitative research examining people at risk for falling have studied sit-to-stand or sit-to-walk movements from armless, backless chairs of a single height, frequently adjusted to 100% of participant knee height. Movements are almost always constrained, requiring arms to be crossed over the chest or placed on the stomach. Many studies additionally stipulate precise placement of participants' pelvic landmarks in relation to the seat edge as well as trunk inclination and foot placement on the floor. We chose to avoid all constraints with the exception of requesting participants perform their rise with one foot on each of two force plates placed directly next to the bedside. While this allowed for an infinite variety of sit-to-walk motor strategies to occur, we felt this would provide a more realistic assessment of how patients actually get out of bed and therefore improve ecological validity of the research, much like the sit-to-walk study done by Frykberg et al. [28] examining temporal coordination between stroke patients and controls. Kerr et al. [29] also argued for the validity of an unconstrained approach.

Lastly, most studies have sought to characterize differences in sit-to-walk movement parameters between elderly at risk for falling and those not at risk in order to identify those differences which may influence stability in the fall-prone. Little investigation has been undertaken to examine how manipulating environmental factors might alter kinematic parameters associated with falling.

1.2.1 Sit-to-Walk Movement Characteristics

Sit-to-walk is an everyday motor task and is fundamental for independence. It requires a complicated overlap of postural stability and locomotor sequences initiating a cascade of events which demand mobility. In other words, it is a dynamic movement requiring an individual to transition from sitting to standing to walking. In most healthy populations, however, it has been found that gait initiation actually occurs before full body extension is complete, unlike sit-to-stand (which obviously terminates with no gait component), and thus walking actually begins within a transition period while an individual is still rising, usually around the seat-off event. This entails moving the center of mass (CoM) up and over a reduced base of support (BoS) while requiring maximal moment generation in the lower extremities. Thus, sit-to-walk (STW) can be considered a more complex motor task than sit-to-stand (STS) with greater demands on stability. It has also been far less studied [30-32].

The STW task has been divided into several generally accepted phases modeled on STS and marked by a variety of events. Many studies diverge on the methodology used to quantify the events, often dependent on their specific study design. However, five phases are typically recognized and used in STW evaluation, first defined and validated by Kerr et al. [29, 33] a decade after Magnan et al. [34] pioneered the key events within the movement and presented evidence for its argument as a single continuous activity vs. a linked combination of STS and gait initiation movements.

The standard four phases are illustrated in Figure 1 and their general features are:

- Phase 1: Flexion momentum – initiation of trunk and lower extremity joint flexion wherein the CoM moves forward and down, generating increasing horizontal velocity.
- Phase 2: Extension – initiation of trunk and lower extremity joint extension

in order to initiate seat-off and rising action. Vertical velocity starts to spike and horizontal velocity decreases until just the end of the phase, after which it begins to reverse.

- Phase 3: Unloading – the initiation of gait, characterized by the swing foot heel coming off the ground and ending with actual swing-off (i.e., first toe-off). It is important to note this phase overlaps with phase 2.
- Phase 4: Stance – the transition between swing-off (first toe-off) and stance-off (second toe-off) which creates first single leg stance. Horizontal velocity increases again to a peak and vertical velocity begins to rise and fall as true gait is established.

Magnan's and Kerr's initial works were performed on healthy young populations and concluded that by merging standing and gait initiation at the point of seat-off, an individual is able to take advantage of the inertial properties of both discrete tasks to springboard directly into gait (it is important to remember here that the original definition of gait initiation is movement beginning from quiet standing, not sitting). Eventually additional research by Kerr [32, 33] and Buckley et al. [30] examined STW within elderly populations and found that many individuals do not perform the STW task as a fluid, continuous motion but rather as disjointed separations of the STS and gait initiation components. They also found that elderly at risk of falling have a greater degree of separation in these tasks than healthy elderly and a wider distribution of timing in each phase, all significantly slower. This agreed with a slightly earlier study published by Dion et al. [35] which examined the CoM horizontal momentum fluidity patterns in stroke vs. control subjects during what she called the rise-to-walk task. Her team found a delayed first toe-off in stroke sufferers following seat-off, concluding that this group of subjects allowed more time for complete standing before initiating gait.

Theories for why fall-prone individuals display marked reductions in locomotor ability in this task include psychological contributions like fear of falling and physiological contributions like reduced muscle strength and power, reduced range of motion, compensatory biomechanics secondary to disease pathology, and impaired balance [30, 32, 36].

1.2.1.1 Human Kinematic Challenges: Stability and Mobility

In light of these differences, an important question is if these altered motor strategies improve or reduce stability in those at risk for falling. General consensus has trended toward kinematic parameters emphasizing stability as being "desirable" at the expense of reduced mobility in order to accommodate impairments [30, 32, 36]. However, it begs the question of when more is no longer better? We hypothesized that some kinematic threshold exists which should provide ideal degrees of both stability and mobility and that seat/bed height plays a significant role. To better illustrate the dichotomy between these two descriptors, it is helpful to review the work of Hughes et al. [37] which delineated three motor strategies related to these different ends of the spectrum in elderly performing STS:

1. **Momentum transfer:** the use of horizontal momentum generated in the trunk to assist in rising via transference to the lower body. Typically

a “fast” movement.

2. **Stabilization:** also called the “zero momentum” strategy and based on repositioning the CoM and the BoS in relation to each other to prioritize stability before rising. Typically a “slower” movement. Figure 2 illustrates this.
3. **Combined:** employing elements of both as needed.

However, none of these results were correlated with fall risks, although two years later the same team examined how these strategies changed during progressively lower seat heights in moderately impaired elderly to investigate reasons behind failed STS. Since failures may be related to fall risk, this type of study provides a little more insight into biomechanical variables of interest. They found that participants increased their trunk momentum at lower seat heights but also tried to increase stability simultaneously, a motor strategy not necessarily seen in healthy populations. In other words, fallers increased their trunk velocity at these low seat heights yet their time to rise also increased. The extra time was not the sole result of the extra vertical distance traveled. Rather participants spent the extra time repositioning their CoM over their BoS prior to the flexion momentum phase and later used additional time to employ stability tactics during the extension phase. The researchers theorized that these two strategies are at odds and thus contribute to failures since increasing momentum generation is not conducive to postural stability [38]. Since then, few researchers investigating STW tasks have directly referenced these strategy classifications with the exception of Chen et al. [39, 40].

1.2.1.2 Variables of Significance

While the literature still remains lacking in the areas of rising failures and how they relate to fall risk, more empirical investigation has been devoted to examining components of stability during STW and a few variables seem to emerge as significant within fallers. These include:

- A significant drop in linear CoM velocity in the anterior direction between seat-off and swing-off events of up to 50%. This is in comparison with a 15-35% drop in healthy populations of all ages [31, 33, 36, 40, 41].
- A long time delay in the duration of the overall task, in particular during the extension phase, which coincides with seat-off and gait initiation, indicating reduced fluidity and mobility [28, 31, 33, 35, 39].
- A center of gravity (CoG) which is positioned more directly over the BoS prior to seat-off. Young and elderly individuals who are healthy typically maintain a CoG which is behind the BoS during this event and transfer over only once they begin rising [34, 38-40, 42, 43].
- A large reduction in CoM linear momentum in the anterior direction during all STW phases, but in particular during the flexion momentum and extension phases when compared to healthy populations. Since momentum generation is a key trait required for successful STW and STS, impaired elderly demonstrating this phenomenon have been termed as using a “zero momentum” strategy for transferring from sitting to rising in some studies [29, 30, 35, 39, 40, 42].

- Generally less energetic preparation and rising phases (phases 1 and 2 in Figure 1) which appears to potentially allow momentum to be mis-translated into the posterior and mediolateral directions as gait initiation occurs, inducing more whole body sway [36, 40, 42].
- Increased mediolateral CoM velocity as gait becomes established. [36, 40, 44].
- A reduced initial step length [31, 36, 39, 40].

1.2.2 Research Aims

The aim of this research was to investigate effects of hospital bed height and side rail configurations on fall parameters in those with a variety of risk factors. Specifically, we wished to examine how these two extrinsic variables affect fall potential as it interacts with a patient's strength capabilities and degree of impairment. We also wished to determine the optimal bed height and side rail configuration that might reduce the risk of injury and provide a fast and novel way for clinical workers to set in-patient bed heights for individuals at risk for falling. To that end, we chose to examine three bed heights in relation to subjects' tibial plateau heights: one measurement considered "low," one measurement considered "medium," and one measurement considered "high" (more detail on the calculation of each will be provided later).

It was hypothesized that the medium bed height would provide the ideal middle ground between creating improved mobility for fall-prone individuals during STW while also allowing them to also use optimal techniques for balance.

It was believed this would occur by allowing “just enough” whole body momentum generation to accomplish successful rising and gait initiation while maintaining postural stability, thus reducing fall risk. To this end, analysis of anteroposterior and mediolateral center of mass velocity and momentum changes between seat-off and toe-off events was deemed important to reveal perturbations in stability between the two axes. It was also hypothesized that the overall task duration as well as the durations occurring between key events of the STW task would be reduced in the bed height providing the ideal bridge between stability and mobility.

This information will provide new insights into fall research in an area that continues to be lacking. It will help evaluate qualitative inquiries, beliefs, and assumptions about hospital fall risks through a quantitative lens and potentially contribute toward shaping policies and practices which may solve part of the complex puzzle of why falls happen and how to prevent them. Lastly, it will add to the groundwork of biomechanical research focusing on the challenges of the STW movement in elderly and impaired populations.

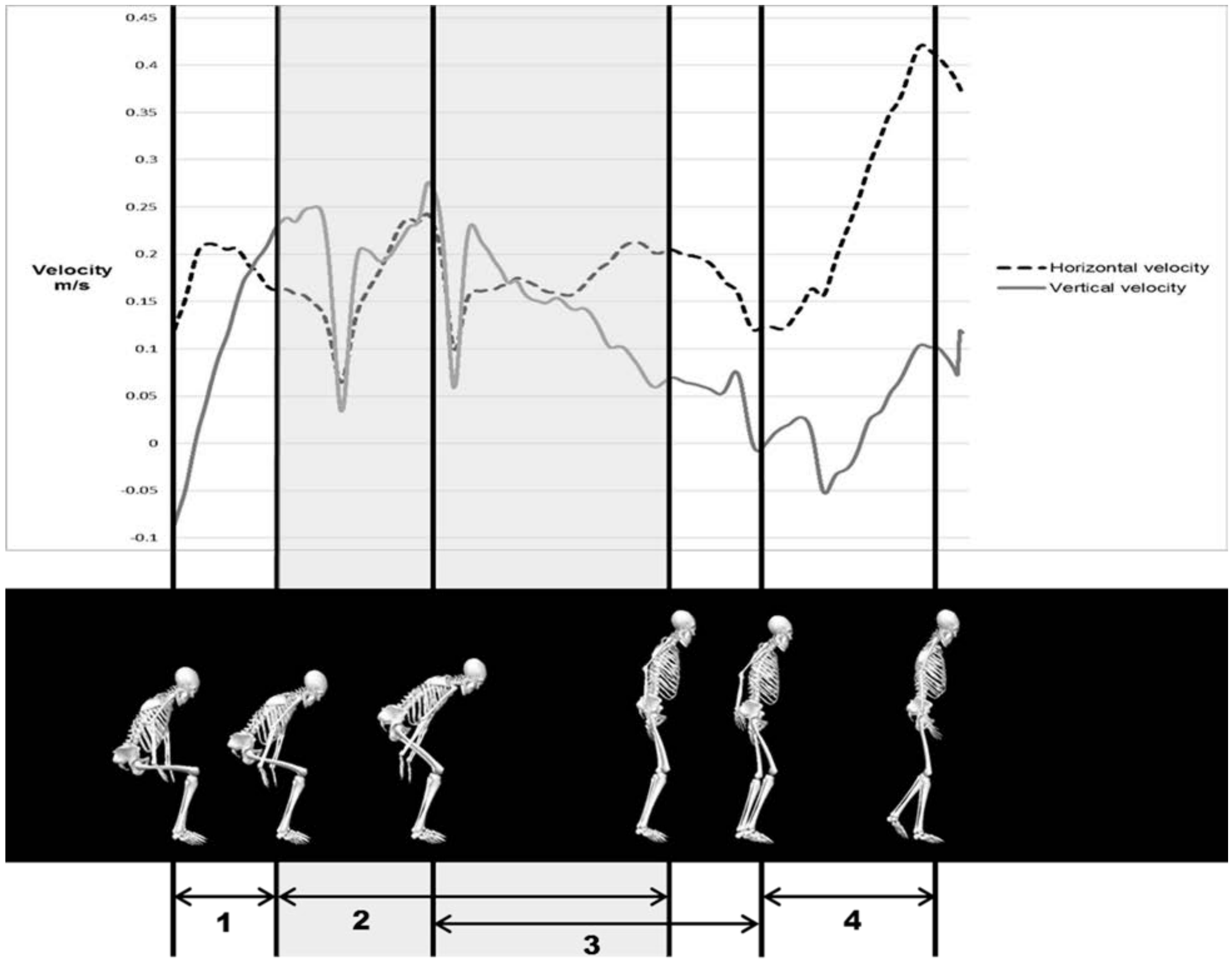


Figure 1. Phases of sit-to-walk in one fall-prone individual:
The gray area between phases 2 and 3 indicates the temporal timing range possible for GI.

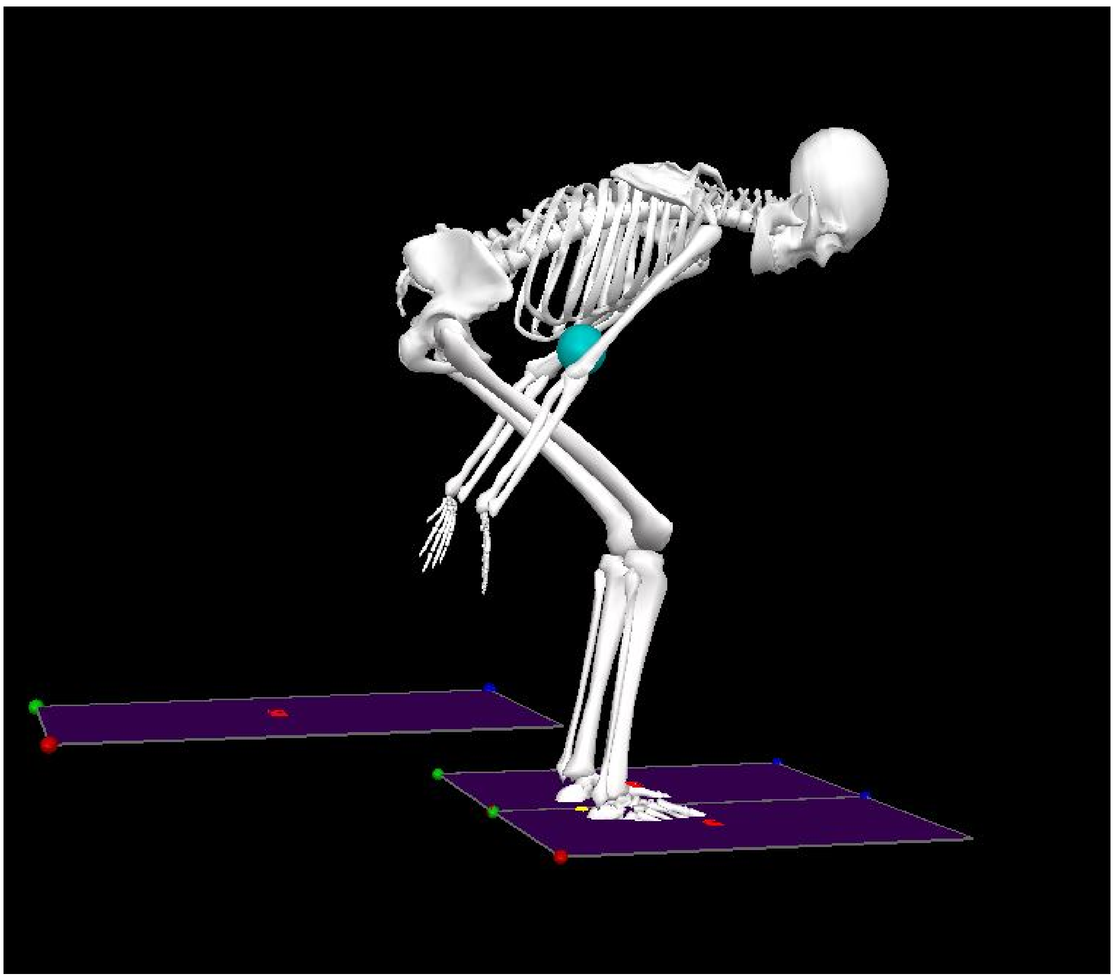


Figure 2. Example of a stability-driven STW strategy.

CHAPTER 2

METHODS

2.1 Participants

Seventy-nine older adults with strength, gait, and mobility impairments were recruited for this study. Participants in this study were a subset of a larger study examining multiple effects of hospital bed height on mobility and stability parameters of fall-prone populations: mean (±SD) age of 69.2±11.0 years, height of 172±10 cm, body mass of 87.3±20.2 kg, and body mass index of 29.6±6.4 kg/m^2. Fifty-four participants were male and 21 were female. Almost 90% reported themselves as Caucasian, with the remaining 10% distributed between American Indian, Native Hawaiian, African American, and unknown/refused. All participants had moderate to high fall risk due to a variety of conditions including but not limited to: Parkinson's, rheumatoid and osteoarthritis, degenerative joint disease, and neurologic deficits secondary to mechanical injury or biological disease.

Inclusion criteria were:

1) Morse Fall Scale score ≥ 45

2) Weak or impaired gait or impaired mobility during sit-to-stand and stand-to-sit
3) Able to transition between sitting and standing at the bedside, turn 180°, and walk several steps without assistance at any point

Exclusion criteria were:

1) Limb amputation
2) Anthropometric and/or medical conditions which could preclude the use of the fall arrest system: height > 200 cm; advanced osteoporosis
3) Unilateral strength deficits > 50%
4) Cognitive impairments which would preclude giving informed consent or following simple instructions to perform bed entry and egress

All participants gave informed consent approved by the University of Utah Institutional Review Board which included an overview of the risks associated with participation. Additionally, they were told they could halt the study procedures and withdraw at any time. All participants were offered a small compensatory fee for their time. Recruitment was carried out within the local Veterans Affairs Medical Center (VAMC) via convenience sampling of patient units and outpatient clinics as well as at the Faint and Fall Clinic in the University of Utah Health Science Center.

Intake forms were completed by participants with staff guidance and included basic demographic data, general medical history that focused on key attributes associated with fall risks, a fall history, and an evaluation using the Morse Fall Scale (Appendix B). This allowed the participant's MFS score to be computed and recorded.

2.2 Capture Space Design and Equipment

The study was carried out at the Nurses Education Laboratory at the local VAMC. The space contained an instrumented, adjustable-height hospital bed which could be fitted with or without side rails. A standard, fixed-height chair with arm rests and a deep seat pan was situated several feet opposite the bedside and used as the originating point for ambulation to and from the bed.

The bed itself was instrumented with a custom force plate (Bertec BP4060, Columbus, OH) capable of detecting and recording side rail usage (pulling and pushing forces with upper extremities). Two additional force plates (Bertec BP4060, Columbus, OH) were installed on the floor directly beneath the bed side in order to collect bilateral lower extremity ground reaction forces at a sampling rate of 500 Hz. Kinematic data were captured using 18 optoelectric cameras (NaturalPoint, Corvallis, OR) mounted to a custom-built metal frame surrounding the bed and chair (16’ x 15’ x 8.5’) and sampled at a rate of 100 Hz. Additionally, a regular video recording camera recorded the entire capture space continuously during each participant trial from a stationary mount. Figure 3 and Figure 4 show the detailed set-up.

Isometric strength was collected with an ErgoFet dynamometer (Hoggan Health Industries, Salt Lake City, UT) for triceps and quadriceps muscle groups. Grip strength was collected using a Jamar hydraulic dynamometer (Patterson Medical, Warrenville, IL).

Additionally, due the high fall risk associated with our recruited population, it was necessary to implement a fall protection system during data collection. Participants wore a rehabilitation fall protection harness which was attached via a locking carabiner to a static 8 mm cord tied with a nonslip knot. Two harness sizes were available, depending on the individual's height and weight. The belay cord was secured by establishing a "toprope" over a lengthwise beam of the metal frame surrounding the capture space and a manual belay was set up with a research staff member (Figure 5 and Figure 6).

A space outside the capture area was created to perform the Timed Up and Go test, a procedure which is widely used for mobility assessment and fall risk screening. The allotted space allowed for the requisite 3 meter walkway and a separate armchair which served as the starting point of the test.

2.3 Data Acquisition

2.3.1 Attire and Markers

Participants were provided with clothing for the study. Garments consisted of tight-fitting "bike style" shorts and tight-fitting tank tops available in unisex sizes from small to XXL. Low cut, snug-fitting socks were also provided with a light rubberized tread pattern on the bottom to reduce slip potential and mimic footwear often worn in a hospital setting.

Three-dimensional kinematic data were captured using retroreflective markers on the following anatomical landmarks (all bilateral except in cases where only a single joint or anatomical landmark exists): supraorbital ridges, lateral skull above the ears, occiputs, the C7 spinous process, midclavicles, sternoclavicular joint, acromioclavicular joints, medial and lateral elbow epicondyles, medial and lateral wrist joint centers, first and fifth metacarpals, the T12 spinous process, anterior superior iliac spines, lateral crests of the iliac spines, posterior superior iliac spines, midsacrum, greater trochanters, medial and lateral knee joint centers, medial and lateral malleoli, calcanei, first and fifth metatarsals, and the midtalar dorsum of each foot. Also, marker clusters were secured to the upper arms, forearms, thighs, and shanks for additional tracking reference. Figure 7 illustrates a participant static capture trial with only markers visible. See Appendix A for detailed marker schematic.

2.3.2 Independent Variables

Our independent variables consisted of three bed heights determined as a percentage of lower leg length (LLL):

- 95% of LLL was considered a low bed (LB)
- 110% of LLL was considered a medium bed (MB)
- 125% of LLL was considered a high bed (HB)

Also, three side rail configurations were used: 1) no side rail present; 2) a Stryker® brand high profile, sturdy side rail; and 3) a Hill Rom® brand low profile side rail which allowed slightly more “play” in its fixture. Retroreflective markers were added to the bed’s four corners to define its physical boundaries. All side rails had three retroreflective markers added to each piece as well.

Each participant completed 27 trials; this total included three independent biomechanical movement trials for each of the nine bed height-side rail configurations. Biomechanical trials recorded were as follows: 1) walk-to-sit movements performed for hospital bed entry; 2) in-bed side-to-side turning from a supine position; and 3) sit-to-walk movements performed for hospital bed exit (which is the select focus of this study). Biomechanical trials were performed in the above numeric order for a logical, continuous capture sequence per bed height-side rail configuration. In order to control for fatigue, bed height-side rail combinations were randomized.

To better illustrate the sequences, the following algorithm was randomized for a total of 27 trials:

(3 fixed order biomechanical trials[†]) *x*

(9 combinations of bed height[††] and side rail[†††])[1]

= 27 unique motion capture trials

2.3.3 Nonbiomechanical Collection Methods

Prior to subject markering and motion capture, participants performed the standardized Timed Up and Go test in their street clothes and shoes. Participants were then asked to don study clothing in order to take anthropometric measurements which consisted of height, weight, sagittal plane chest depth between the xyphoid process and the T10 spinous process, frontal plane pelvic width between the anterior superior iliac spines (in cases of high obesity levels),

[1] [†] entry, turning, exit
[††] high, medium, low
[†††] Stryker®, Hill Rom®, none

waist circumference, calf circumference, thigh circumference, and lower leg length. All bone breadths were measured with an anthropometric beam caliper. All circumferences and limb lengths were measured with a calibrated, flexible plastic tape measure. Waist circumference was recorded from the widest part of the lower trunk, and leg circumferences were recorded from the widest part of the muscle bellies. Lower leg length was measured from the heel to the top of the head of the fibula below the lateral aspect of the knee. This is also referred to as the tibial plateau.

During strength data collection, participants were instructed to exert maximal possible force for a 4-second time period for all muscle groups. Two trials each for data acquisition were performed with a minimum 3-minute rest to reduce effects of fatigue. All strength data were collected while the participant was seated. Triceps collection required the individual to align their forearm on the armrest next to the body with the shoulder at 0°, the elbow at 90°, and the shoulder nonelevated. A circular strap attached to a chain and connected to the load cell was placed around the wrist and the individual was instructed to push down as hard as possible without moving the arm or shoulder. Grip collection was in the same position but simply performed by squeezing the dynamometer. Quadriceps collection required the individual to maintain the hip and knee at 90° angles and to minimize motion of the upper leg. The circular strap was placed around the ankle and the individual was instructed to push the ankle forward as hard as possible without moving the upper leg or trunk.

2.3.4 Biomechanical Collection Methods

Following the above measurements, the safety harness was put on the participant and the markers were placed on the body. The lower leg length was used to calculate the bed height percentages and then the trial order was randomized. After the order was determined, it was noted on a whiteboard and the first trial height was set by measuring the distance from the floor to the deck height of the bed at a perpendicular angle using a yardstick. This was chosen vs. measuring to the mattress top in order to control for mattress compression under body weight while sitting on the edge of the bed.

Before collecting trial-specific data, a 30-second static capture was performed. The participant was first put on belay by a research staff member via the static cord attached to their chest harness and then asked to stand up, walk toward the bed and stop once their feet were centered on each of the two force plates. They then raised their arms straight out to approximately shoulder height with palms down, kept their heads lifted and level, and remained motionless for the 30-second duration (Figure 6).

Once the static pose was captured, medial ankle and knee markers were typically removed since they often posed a problem of appearing to merge in the optoelectric view field (due to marker proximity during walking and sitting). Once this was accomplished, the randomized trials were set to begin.

Each entry trial consisted of the participant rising from the chair, walking to the bed, turning, and sitting down. All participants were allowed to walk and turn at a self-selected pace and direction and no instructions were given regarding side rail usage (when present).

All turning trials began with the participant in the sitting position assumed at the end of the entry trial. They then moved into a supine position centered on the bed, rolled onto their left sides, returned to supine, rolled to their right sides, returned to supine, and then sat back up on the side of the bed from which they started. In some cases of highly impaired participants, turning trials were truncated for safety reasons. Truncated trials typically occurred in no-side rail bed conditions and in all mid and high bed height conditions, with or without side rails. These trials were chosen for exclusion in light of safety considerations (i.e., higher bed height and lack of rail guarding). A truncated trial typically consisted of a participant either lying supine for several seconds then returning to a sitting position or simply not performing the trial at all.

Each exit trial consisted of the participant rising from the side of the bed, walking to the chair, turning, and sitting down. In the case of exit trials, no constraints were placed on participants' sit-to-walk style with the exception of the request that they place one foot on each force plate prior to rising. Otherwise, use of upper extremities, foot placement, speed, and motor strategy were self-selected.

2.4 Data Analysis

2.4.1 Creation and Processing

The biomechanical variables chosen for analysis were created using temporal, kinematic, and kinetic data collected and synchronized between the three-dimensional motion capture and the ground reaction force. Analog data were filtered using a digital lowpass filter (Butterworth or critically damped) available within Visual3D software (C-Motion Inc., Germantown, MD) with a cut-off frequency of 15 Hz. Motion data were filtered using the same filter with a cut-off frequency of 6 Hz. A custom whole body model was created to provide robust modeling across hundreds of trial types, body phenotypes, and physical motor characteristics. The model consisted of 15 linked segments with representative inertial characteristics. Figure 8 and Figure 9 illustrate the link model and the lab origin referenced.

2.4.1.1 Events

Identical events were created in every trial considered for analysis. Variables of interest were extracted from the data provided by the kinematic, kinetic, and temporal measurements occurring at the events of note. Events considered important were related to STW phases and recorded at their respective frame numbers *"i."* In turn, frame numbers signified corresponding time values by the following equation:

$$i = \frac{t}{100}\, sec \qquad \text{(Eq. 1)}$$

All equations in this text involving derivatives or other mathematical markers of time can also substitute the formula "*i**100" in place of *t.*

Event descriptions and calculations are as follows.

- Movement initiation: defined as the first detected displacement of the trunk CoM in the Y direction using kinematic data; the threshold was triggered by taking the average trunk CoM movement on the Y axis in the first few frames of the trial and adding two standard deviations beginning at frame number one. The identifying frame number of this event was used to mark t_0 for all temporal parameters calculated later.

$$t_0 = MOVE\_INT = \frac{1}{8}\sum_{i=1}^{8} y_i + 2\sqrt{s^2} \qquad \text{(Eq. 2)}$$

where $$s^2 = \frac{1}{(8-1)}\sum_{i=1}^{8}(y_i - \overline{y})^2$$

- Begin to stand: defined as the first detected displacement of the trunk CoM in the Z direction using kinematic data; the threshold was triggered by taking the average trunk CoM movement on the Z axis in the first few frames of the trial and adding two standard deviations starting at MOVE_INT. This event typically coincides with the beginning of the flexion momentum phase.

$$BEGIN\_STAND = \frac{1}{8}\sum_{i=t_0}^{8} z_i + 2\sqrt{s^2} \qquad \text{(Eq. 3)}$$

where $$s^2 = \frac{1}{(8-1)}\sum_{i=t_0}^{8}(z_i - \overline{z})^2$$

- Peak anteroposterior ground reaction force: created using the maximum value of the force plate output signal on the Y axis in the fixed reference frame “N.” This kinetic event typically coincides with seat-off in the STW movement and the beginning of the extension phase. See Figure 10.

$$PEAK\_AP\_GRF = i\ when\ \frac{d}{dt}|_N\ \overrightarrow{GRF_y}(t) = 0 \qquad \text{(Eq. 4)}$$

- Peak CoM velocity in the Z direction: created through differentiating the Z position vector of the body's CoM in the fixed reference frame "N" and finding the first local maximum value of the signal after BEGIN_STAND. In healthy populations, this often defines the end of the extension phase and overlaps with the unloading phase.

$$PEAK\_COM\_VEL = i\ when\ \frac{d}{dt}|_N\ \vec{r}_z(t) = 0 \qquad \text{(Eq. 5)}$$

- Heel-off: created by marking the instant when the first of two heel targets crossed a threshold of 0.02 m/s in the positive Z direction of the fixed reference frame "N" following the peak CoM velocity event. Positive signal slope was determined using code syntax which identified whether a slope was ascending or descending.

$$HO = i\ when\ \frac{d}{dt}|_N\ \vec{r}_z(t) > 0.02\ m/s \qquad \text{(Eq. 6)}$$

- Swing-off/first toe-off: all toe-offs were created by referencing the first ray metatarsalphalangeal marker (i.e., RTOE or LTOE) positions from the fixed reference frame "N" to the rotating reference frame "P" of the pelvis segment CoM. A frame was recorded as a toe-off event when the toe marker's signal was detected to be at a minimum value on the Y axis relative to the reference frame "P." Extrema were identified using code syntax which determined slope sign (i.e., minima was captured when slope was negative). Swing-off was considered the first instance of this event of the right or left toe marker following heel-off. The transformation coordinate math used to obtain the signal minimum was:

$$If\ \vec{r}_{toe} = \begin{bmatrix} r_y \\ r_x \\ r_z \end{bmatrix}_N \ and\ \ \vec{r}_{toe} = \begin{bmatrix} r_{y'} \\ r_{x'} \\ r_{z'} \end{bmatrix}_P$$

$$then \begin{bmatrix} r_{y'} \\ r_{x'} \\ r_{z'} \end{bmatrix}_P = T \begin{bmatrix} r_y \\ r_x \\ r_z \end{bmatrix}_N \ where\ T = \begin{bmatrix} -\cos\theta & \sin\theta & 0 \\ \sin\theta & -\cos\theta & 0 \\ 0 & 0 & 1 \end{bmatrix} \qquad \text{(Eq. 7)}$$

$$\therefore TO = i\ when\ \frac{d}{dt}|_P\ \vec{r}_{y'}(t) = 0$$

- Stance-off/second toe-off: created identically to swing-off but considered the second occurrence of toe-off following heel-off.

<u>2.4.1.2 Parameter Signals</u>

Velocity parameters were measured as instantaneous outputs derived from local body position vectors at events. Linear components were extracted and used independently. The pelvis segment CoM was used as a surrogate for whole body CoM due to its proximity. The trunk segment CoM was used as a measure of whole body momentum (Figure 11). Local kinematic parameters were calculated using relative motion equations and the requisite transformation coordinates from the fixed reference frame “N” to the segment rotating reference frame “S.”

$$If\ \vec{v}_{segment} = \begin{bmatrix} v_y \\ v_x \\ v_z \end{bmatrix}_N \ and\ \ \vec{v}_{segment} = \begin{bmatrix} v_{y'} \\ v_{x'} \\ v_{z'} \end{bmatrix}_S \qquad \text{(Eq. 8)}$$

$$then \begin{bmatrix} v_{y'} \\ v_{x'} \\ v_{z'} \end{bmatrix}_S = T \begin{bmatrix} v_y \\ v_x \\ v_z \end{bmatrix}_N \ where\ T = \begin{bmatrix} -\cos\theta & \sin\theta & 0 \\ \sin\theta & -\cos\theta & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

All events were coded and then verified individually by the research team in order to catch errors and adjust actual event timing when necessary. Data were then exported and analyzed using MATLAB (The Mathworks, Natick, MA) and SPSS Statistics 22 (IBM Corp., Armonk, NY).

2.4.1.3 Error Checking

Once exported, data plots were created in MATLAB to examine outliers and check for abnormalities. Each file was individually reviewed; if errors were present, they were either fixed or the trial was excluded. Additionally, code was generated to check for presence of replicated events within single trials as well as event sequencing errors, such as right or left toe-offs occurring sequentially. These trials were also individually checked and errors were corrected when possible. Once all outliers and errors were reviewed and updated, all signals were re-exported and included in the working data set. This resulted in 689 unique trials out of 738 possible. An additional step of filtering unusable data was done in MATLAB. Trials were excluded from analysis when they displayed force errors, marker errors, missing body segments, or were incomplete. Trials displaying these errors were originally tagged with the correlating identifiers in Visual3D; MATLAB code was generated to exclude these tagged trials from any statistical testing. This resulted in the loss of 32 observations: 9 in the no-side rail condition, 11 in the Stryker® side rail condition, and 12 in the Hill Rom® side rail condition.

2.5 Statistical Analysis

Statistical power for the study was calculated during data collection by the professional statistics team employed by the grant. This allowed the research team to more accurately determine additional participant numbers required. Descriptive statistics were calculated for population characteristics (Table 1). Two-way analysis of variance (ANOVA) was run on each dependent variable (DV) to determine significance due to bed height, side rail condition, and their interaction. Tukey-Kramer pairwise comparisons were run for the three bed heights and the three side rail conditions to determine statistically significant differences between groups for each dependent variable. This pairwise comparison allowed for reduction of type I error and to allow for unequal numbers of observations between groups. Results were considered significant when $p < 0.05$ ($\alpha = 0.05$).

Repeated measures analysis of variance (RM ANOVA) was used to determine the effects of each bed height on STW movements in the no-side rail condition. Cut-off for significance was maintained at $p < 0.05$ ($\alpha = 0.05$). When the assumption of sphericity was violated, the Greenhouse-Geisser correction was used when $\varepsilon < 0.75$; otherwise the Huynh-Feldt correction was used. Post-hoc pairwise comparisons were conducted using the Sidak correction factor to reduce type I error but maintain power since comparisons were fewer than six. Observed power and effect size using partial eta squared were calculated. Effect size was considered small when $\eta^2_P = 0.01$, medium when $\eta^2_P = 0.09$, and large when $\eta^2_P = 0.25$ per Cohen's d guidelines. It was calculated using the following equation:

$$\eta_P^2 = SS_{Effect}/SS_{Error(Effect)} \qquad \text{(Eq. 9)}$$

When considered relevant, RM ANOVA was performed on subsets of the data.

Principal Component Analysis was also run to determine which singular dependent variables could represent the maximum variation of the data set and thereby omit redundant measures.

To examine effects of covariates as IVs, stepwise regression analysis was performed on two DVs occurring at crucial events bookending the dynamic transition phase of STW: time to peak A/P GRF (seat-off) and time to swing-off. Covariates included age, MFS score, TUG score, BMI, and maximal isometric strength of the right quadriceps. Additionally, bed height was included as an IV.

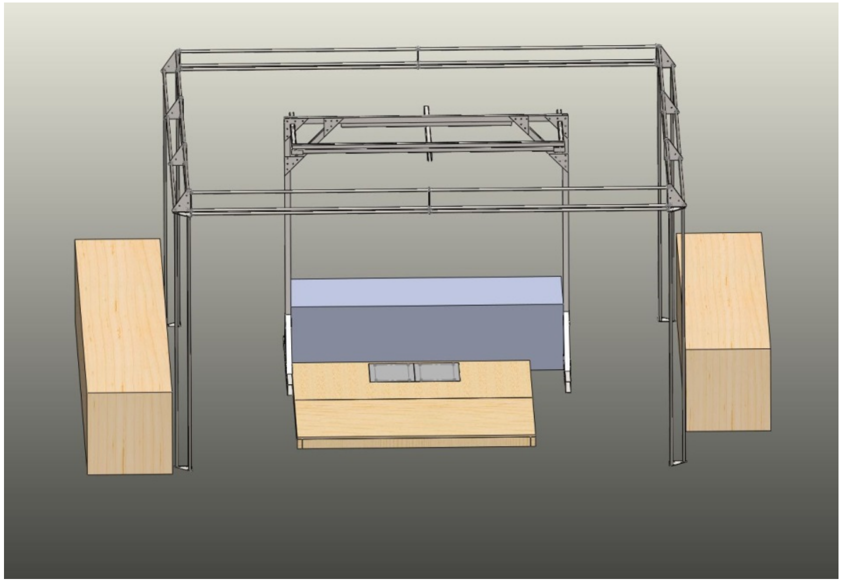


Figure 3. Capture space design.

Figure 4. Overview of the capture space with the fall arrest system removed.

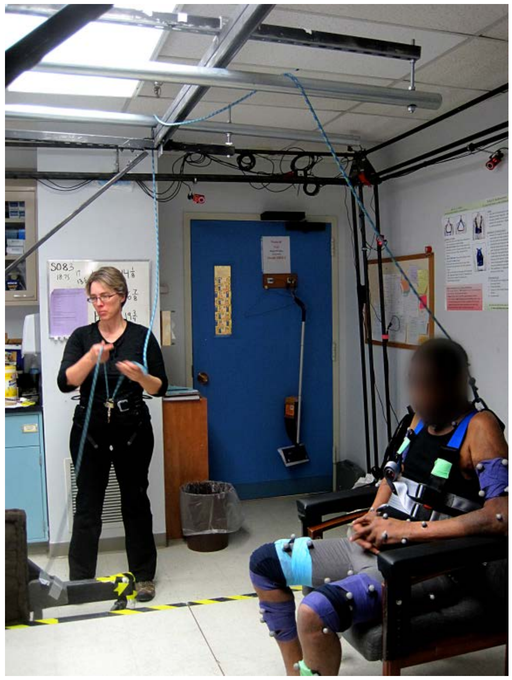


Figure 5. Fall arrest system.

Figure 6. Static capture on force plates.

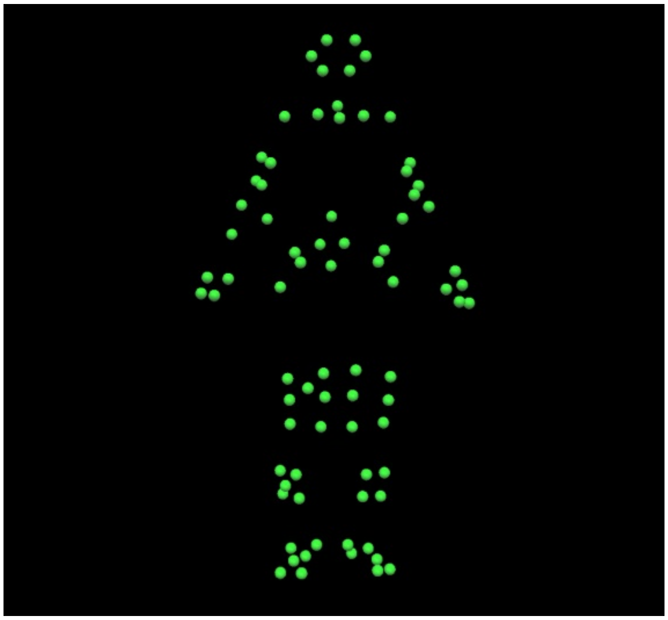


Figure 7. Static capture highlighting only markers.

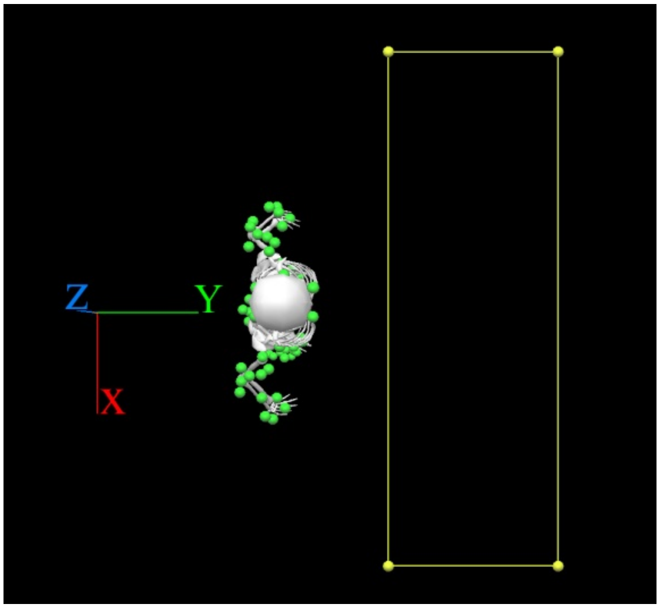


Figure 8. Overhead view of markered bed, markered participant, and lab axis orientation.

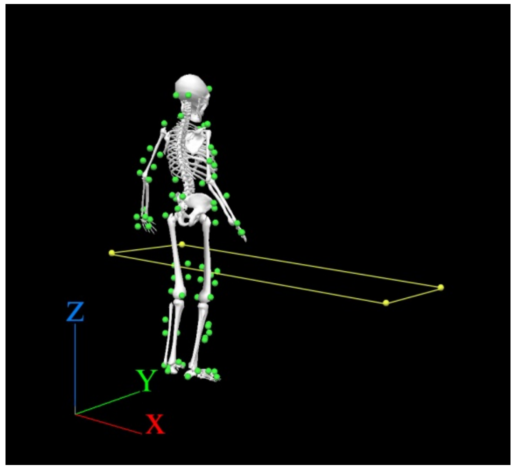


Figure 9. Rear lateral view of markered bed, markered participant, and lab.

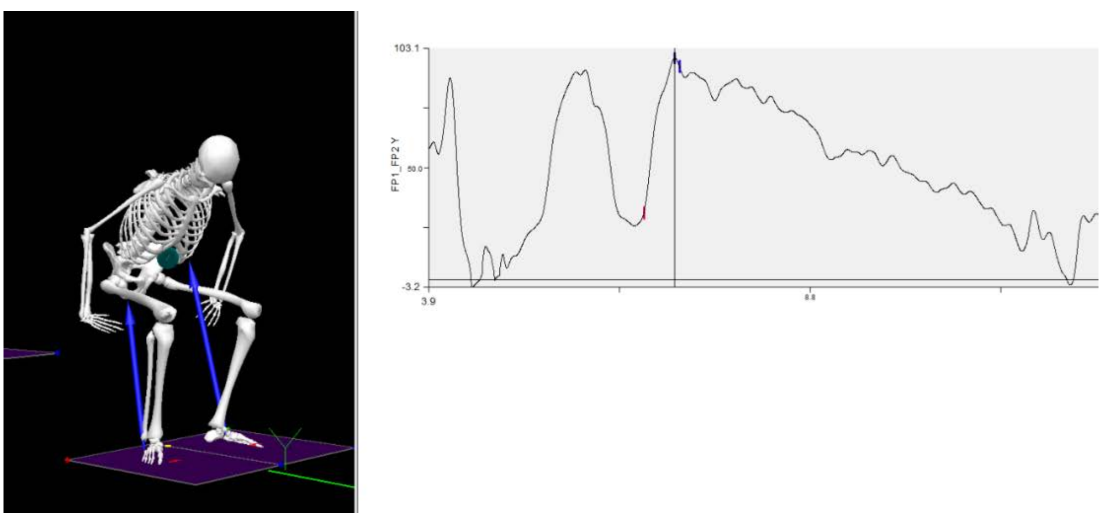


Figure 10. Highlighted peak anteroposterior ground reaction force event and concomitant seat-off.

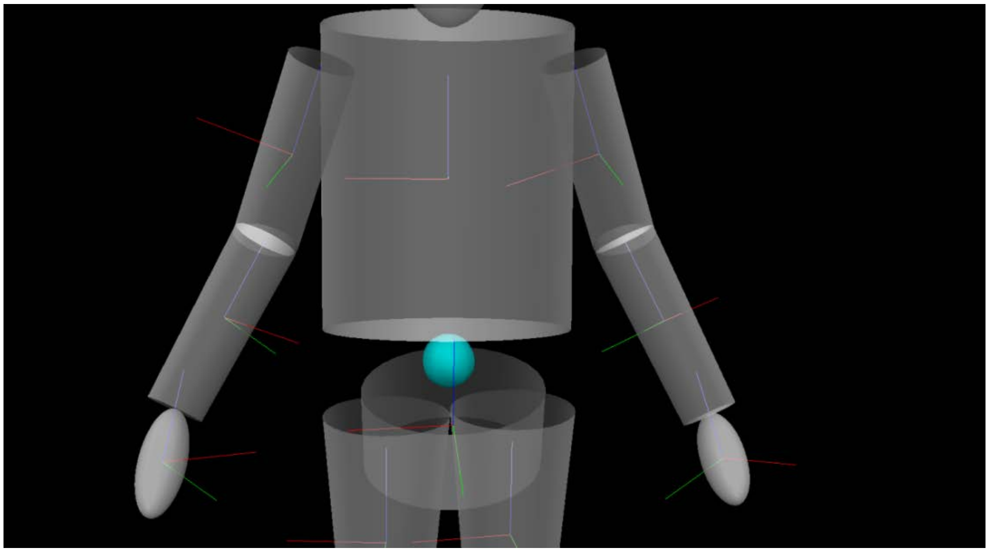


Figure 11. Close up of the pelvic and trunk segment CoM coordinate systems used to produce local kinematics.

Table 1. Participant Demographics by Morse Fall Scale

	Total		MFS ≤ 50		MFS > 50	
Gender	N	%	N	%	N	%
Male	55	69	27	66	28	70
Female	25	31	13	32	12	30
Characteristics	Mean	SD	Mean	SD	Mean	SD
Age (yrs)	69	11	66	14	71	11
BMI (kg/m^2)	30	6	28	7	30	7
MFS score	54	21	37	13	72	11
TUG (s)	15	7	13	6	18	7
Height (cm)	171	10	167	27	171	10
Weight (kg)	87	20	82	23	88	20
Lower Leg Length (cm)	44	3	43	7	45	3

CHAPTER 3

RESULTS

3.1 Bed Height and Side Rail Conditions

Preliminary two-way ANOVA results indicated the presence of side rails did not influence human kinematics during STW movements and there was no interaction between the side rail and bed height. Furthermore, in most exit trials, the majority of participants did not use the side rail. At this point, the data set was recoded to identify side rail users from nonusers in order to evaluate if kinematic differences existed between them when bed height was a constant. After recoding to identify users, it was apparent that the group numbers were highly unequal and not suitable for comparison since nonuser numbers were triple to quadruple the number of users. Table 2 illustrates the number of usable trials in each condition and the breakdown of side rail use. Analysis was then focused on nonside rail only conditions for each bed height and side rail trials were set aside for separate analysis at a later time.

Principal Component Analysis (PCA) was performed on the final no-side rail condition data set and resulted in 13 eigenvalues greater than one. A scree plot indicated a cut-off point which included 7 component groups of variables and the variables' respective correlations within each component axis. Variables were ordered according to their degree of correlative value. One variable from each group was chosen, usually with the highest correlation to the component, to represent all other similar items within the component and thereby reduce the dimensions for analysis to seven representations. Table 3 displays the component groupings and the variables chosen, their Eigenvalues, and the percent variance of the total data set for which each component accounted.

3.2 Bed Height as a Singular Predictor

Once it was determined that bed height played the primary role in influencing STW kinematics, RM ANOVA was performed on no-side-rail-only trials. This method was chosen because it would yield a data set in which individuals performed STW at all three bed heights yet would avoid learning effects by adding observations of the same individuals in unequal numbers across conditions. This data set required the exclusion of 14 participants because they lacked three complete no-side-rail trials. This changed the descriptive statistics minimally. Table 4 summarizes the following sections 3.2.1-3.2.2.

3.2.1 Temporal Variables

Bed height was found to significantly influence the timing of several variables including total task time ($p=0.016$), the extension phase duration ($p=0.001$), the ratio of the extension phase normalized to total task time, aka time ratio ($p=0.006$), the heel-off time, the swing-off time, and the stance-off time (all p=0.006).

3.2.2 Velocity Variables

3.2.2.1 Anteroposterior

Bed height was found to significantly influence the change in velocity between peak anteroposterior GRF (i.e., seat-off in most participants) and swing-off (p=0.011). P-values were between 0.05 and 0.10 for instantaneous velocity at the swing-off event (p=0.067).

3.2.3 Momentum Variables

3.2.3.1 Anteroposterior

Momentum occurring at peak A/P GRF was found to be significant between bed heights (p=0.047) although post-hoc pairwise comparisons did not support this result (additional detail will be provided in the discussion). A/P momentum at peak vertical velocity was also significantly different between bed heights (p=0.000).

3.2.3.2 Mediolateral

Stance-off momentum was found to be statistically significant between bed heights (p=0.042) but post-hoc comparisons did not bear this out. Additional detail will be provided in the discussion.

3.3 Regression Models

3.3.1 Time to Peak Anteroposterior Ground Reaction Force

Linear regression resulted in two models using predictors:

1) TUG only (p=0.000)

2) TUG plus low bed height (p=0.000)

The adjusted R^2 was .250 for model 2; the addition of low bed height accounted for 0.023 of that total. Model 2 can be mathematically represented as:

$$Time\ to\ peak\ A/P\ GRF = 0.041 + 0.105x_1 + 0.463x_2\ sec$$

where $x_1 = TUG\ score$ (Eq. 10)

$x_2 = Presence\ of\ low\ bed\ height$

3.3.2 Time to Swing-off

Linear regression resulted in three models using predictors:

1) TUG only (p=0.000)

2) TUG plus low bed height (p=0.000)

3) TUG plus low bed height plus MFS (p=0.015)

The adjusted R^2 was .334 for model 3; the addition of low bed height and MFS accounted for 0.077 of that total. Model 3 can be mathematically represented as:

$$Time\ to\ swing-off =$$

$$-0.118 + 0.152x_1 + 1.107x_2 + 0.018x_3\ sec \quad \text{(Eq. 11)}$$

where $x_1 = TUG\ score$

$x_2 = Presence\ of\ low\ bed\ height$

$x_3 = MFS\ score$

Table 2. Side Rail Usage Statistics for All Exported Trials.

Side rail condition	**No. of trials**	**No. using side rail**	**% of the total**
None	277	N/A	N/A
Stryker®	230	82	36
Hill Rom®	232	57	25

Table 3. Principal Component Analysis Results.

No.	Component Descriptor	Variable Chosen	Eigenvalue	% Variance Explained	Cumulative Variance
1	Time	Swing-off time	15.3	27.7	27.7
2	A/P General Gait Velocities	Velocity at swing-off	7.9	14.3	42.0
3	M/L First Step Kinematics	Velocity at swing-off	4.3	7.8	49.8
4	M/L Second Step Kinematics	Momentum at stance-off	3.6	6.5	56.3
5	STW Peak Kinematic Values	Peak M/L momentum	3.1	5.7	62.0
6	Rate of Change - Second Step Kinematics	Velocity Δ swing-off to stance-off	2.9	5.2	67.2
7	A/P First Step Momenta	Momentum at heel-off	2.1	3.8	71.0

Table 4. Summary of Repeated Measures ANOVA.
N varies between analyses due to occasional missing data and listwise deletion.

Bed Height Main Effects - Within Subjects

	Condition	Mean(SE)	95% CI	N	F (p)	η^2_P	Power
Temporal							
	H*	3.7(0.3)	3.2-4.2				
Total task time (s)	M*	3.9(0.3)	3.3-4.5	61	7.899(0.006)	0.116	0.950
	L**	6.3(1.0)	4.4-8.2				
	H*	1.3(0.1)	1.0-1.5				
Extension phase time (s)	M*	1.4(0.2)	1.1-1.8	61	10.570(0.001)	0.150	0.988
	L**	2.4(0.4)	1.7-3.1				
	H*	32.4(1.5)	29.4-35.3				
Time ratio %	M	35(1.5)	32.0-38.1	61	6.227(0.006)	0.094	0.812
	L*	38.8(2.3)	34.3-43.3				
	H*	2.6(0.3)	2.3-3.4				
Heel-off time (s)	M*	3.1(0.3)	2.5-3.7	61	7.878(0.006)	0.116	0.798
	L**	5.4(1.0)	3.5-7.3				
	H*	3.1(0.3)	2.5-3.6				
Swing-off time (s)φ	M*	3.3(0.3)	2.7-3.9	61	7.871(0.006)	0.116	0.798
	L**	5.6(1.0)	3.7-7.5				
	H*	3.7(0.3)	3.2-4.2				
Stance-off time (s)	M*	3.9(0.3)	3.3-4.5	61	7.899(0.006)	0.116	0.800
	L**	6.3(1.0)	4.4-8.2				
Velocity and Momentum							
	H	9.9(2.3)	5.2-14.6				
Δ velocity: seat-off to swing-off (+cm/s)	M*	15.1(2.5)	10.0-20.2	65	4.722(0.011)	0.069	0.781
	L*	9.9(2.3)	5.3-14.5				
	H	1.2(0.6)	0.0-2.3				
A/P momentum at seat-off (kg*m/s)†	M	0.3(0.6)	(-)0.8-1.5	64	3.221(0.047)	0.049	0.582
	L	(-)0.9(0.5)	(-)1.1-0.9				
	H**	4.1(0.4)	4.8-3.4				
A/P momentum at peak v_vel (-kg*m/s)	M*	6.2(0.5)	7.2-5.3	65	29.785(0.000)	0.318	1.000
	L*	7.1(0.5)	8.0-6.2				
	H	5.0(0.5)	4.0-6.0				
M/L momentum at stance-off (kg*m/s)†	M	4.8(0.4)	3.9-5.6	65	3.254(0.042)	0.048	0.611
	L	5.5(0.5)	4.5-6.5				

* Indicates p<0.05 compared to one other bed height
** Indicates p<0.05 compared to two other bed heights
† Indicates statistical significance was not found with post-hoc testing
φ Representative variable of PCA component 1

CHAPTER 4

DISCUSSION

As previously stated in the introduction, no known study to date has examined STW kinematics and temporal characteristics as they relate to differing seat heights. The strategy with this research was to create a link between existing hypothesized fall-prone population risk factors and consider the interaction with bed settings in an unfamiliar environment.

4.1 Temporal Variables

Most temporal variables of significance revealed a performance schism between LB and that of HB and MB conditions. All effect sizes for temporal variables fell between the range of 0.09 to 0.25 and are considered medium by Cohen's d cut-offs. Observed power was good for all variables, being 0.8 or higher. In the case of total task time, pairwise comparisons revealed LB to be significantly different from HB (p=0.024) and MB (p=0.016). Participants took, on average, 2.4 seconds longer to complete the STW task from the LB. These delays in STW execution may mean more postural accommodations to maintain balance were required in the LB condition [33] and are in agreement with data produced by Buckley [31] on Parkinson's patients and Frykberg [28] in stroke patients.

Extension phase durations were also significantly different between LB and HB (p=0.002) and between LB and MB (p=0.013). HB and MB extension phase durations were, on average, 1 second faster than LB. The extension phase characterizes the time spent rising as it overlaps into gait initiation and may be the most demanding transition of the task due to the dynamic use of the body's inertial properties and the need for simultaneous postural control. This study distinguished the end of this phase at physical heel-off, i.e., gait initiation, due to the fact that many older, mobility-impaired individuals do not overlap the two events in the same manner as younger and/or healthier counterparts. Also, longer time-to-rise phases are typically present in fall-prone individuals and may be a hallmark of the need for increased postural adjustments to accommodate a lack of stability and poor motor coordination between lower extremity muscles [28, 31, 45]. If we extrapolate this to the LB time-to-rise performance, we can interpret this to mean that individuals felt the need to deliberately separate their movements into more perfunctory STS and gait initiation event divisions, in agreement with Kerr et al. [33] in which they note it "may reflect the overall lack of ability or confidence in this group" (p.15).

When the study group was reduced into a subset to examine the same statistical measures on only those with a MFS score of 50 or above, pairwise comparisons revealed all bed height combinations to be statistically significant, including HB and MB (p=0.013). However, the mean difference between these two conditions, an increase of 0.3 seconds in MB, may not be clinically significant despite good observed power (0.997) and large effect size (0.393).

It was also important to consider the extension phase duration relative to the total task time in order to normalize this variable between bed heights as a percentage of the entire STW duration. It was found that the time ratio was significant between HB and LB only ($p=0.009$). HB and MB had a p-value between 0.05 and 0.10 and there was no significance between MB and LB. This means that participants spent more time in the extension phase when rising from the LB condition even when considered as a percentage of the total task time. This could indicate HB requires the least posturally challenging transition prior to the start of gait and the best use of generated kinetic energy (in this case, “posturally challenging” indicates challenges to an individual’s ability to effectively maintain balance within the BoS). Alternatively, it could mean that individuals have a harder time controlling the rate of postural overlap between sitting and standing and thus they perform “quickly” by default. The MB condition appeared to offer the most flexibility either way.

Heel-off, swing-off, and stance-off times were all statistically significant between the LB and HB ($p \leq 0.016$ for all three variables) and the LB and MB ($p \leq 0.025$ for all three variables). LB conditions showed the largest delay in gait initiation by an average of 2.4 seconds, a result that is consistent with the total task time increase for the same condition. Increased hesitancy in STW has been hypothesized to reflect an increased risk of falling in older people as it may imply reduced fluidity, difficulties with mobility, or reveal a misdirection (backwards or sideways) of kinetic energy which could precipitate a fall [36, 40, 41]. A bed height which produces the slowest event times may pose the greatest challenge to a patient's fluidity and so influence fall risk.

The regression model (Equation 11) for swing-off, a surrogate measure for all three gait events, supported the RM ANOVA findings that low bed height (vs. high and medium) increased time to event. Specifically the model predicted that the presence of low bed height would add 1.1 seconds to time to swing-off on average, about half the time estimated by RM ANOVA. It also accounted for TUG and MFS scores, predicting that for each 1-second increase in TUG and each 5-point increase in MFS combined, time to swing-off would increase by approximately two-tenths of a second on average (where TUG was the major factor). This seems minute in the context of the math, but if we consider an example of a 30-second TUG score and an MFS score of 75 against a similar 12/40 combination, this predicts an individual will take 5 seconds longer to begin walking, revealing a significant gap during which an individual may be struggling to establish stability. Since individual characteristics such as TUG and MFS scores cannot be altered, it seems plausible to conclude that any outside measure which decreases the time to get up and walk increases an individual's stability. The regression model appears to support this.

Interestingly, no difference in timing of peak A/P GRF was found, even when the data were subsetted to examine only individuals with a MFS score of 50 or higher. Peak A/P GRF has been used to mark seat-off in highly impaired populations [31] and this study found it also approximated the same kinematic event through the trial-by-trial qualitative error-checking performed on the entire data set (Figure 10). Obviously, there were still a number of individuals who created a peak A/P GRF before or after seat-off, although it is not believed the number is high. This result indicates that seat-off timing is identical for any given bed height and the diverging temporal parameters come into play only upon rising.

Despite the lack of statistical significance between bed heights, this event signifies the beginning of the crucial transition phase into standing and walking. The regression model created in Equation 10 to further examine this outcome suggests there is a relationship between the timing of seat-off and bed height as it interacts with the TUG metric. The predictive model indicates that for each additional second added to TUG score, time to seat-off will increase by one-tenth of a second on average. If there is presence of low bed height (vs. medium or high), time to seat-off will increase by one-half second. However, it is possible this may not prove clinically significant when considering actual fall outcomes.

4.2 Velocity and Momentum Variables

Velocity and momentum parameters are frequently analyzed by components in the STS and STW literature. As a direct result, many studies have concluded that fall-prone individuals have higher lateral velocities and momentums than their healthier counterparts [36, 40, 44, 46]. This is theorized to be associated with a lack of postural control and thus to contribute to instability and fall risk. This study found most velocity and momentum parameters to have no significant differences between bed heights. One exception was the change in velocity between peak A/P GRF, aka seat-off, and swing-off. Pairwise comparisons revealed differences in MB and LB (p=0.012) as well as a borderline significance between MB and HB (p=0.057). There was no difference between HB and LB, interestingly, because of the nature of the metric: HB started fast and stayed fast while LB started slow and stayed slow. MB showed the largest increase in velocity between the two events, a hallmark of healthy, mobile individuals which points to an ability to take advantage of the momentum generated by trunk flexion and channel it through the kinetic chain into rising and walking movements [41, 45].

Instantaneous velocity values at STW events were not significant between conditions, although swing-off velocity approached the 0.05 threshold for MB and LB (p=0.061). On average, MB swing-off velocities were slightly faster than either HB or LB with good observed power and small effect size.

Momentum variables found to be significant include A/P values at peak vertical velocity. The HB condition produced markedly different results from both other bed heights (p=0.000 in each case). All trunk momentum values were negative during this event due to the fact that it occurs during rising. HB required less negative momentum at this event, indicating a potential difference in timing and/or a decreased need for forceful trunk extension to rise. This may mean a high bed height could allow more momentum to be channeled into an anterior direction for forward mobility or simply reduce destabilizing posterior momentum. This particular result had extremely high observed power and large effect size although there are virtually no literature sources which investigate kinematic details (beyond the Z axis) occurring at this event.

RM ANOVA revealed statistical significance between bed height for A/P momentum at peak A/P GRF (p=0.047) but post-hoc testing did not bear this out. It is possible this parameter may have been confounded by differing signal signs. HB and MB momentum means were (+) for the event while the LB momentum mean was (-). It appears that most participants were able to seat-off while still flexing the trunk in the HB and MB conditions whereas they delayed seat-off in LB to occur with trunk extension. This may reveal the need for altering extensor muscle moments to rise from the LB. This finding appears to agree with work by Schenkman et al. [47] who, when examining four chair heights at 115, 100, 80, and 65% of knee height, found that:

> On average, when standing up from chairs set at 80 and 65% KH, these older subjects appeared to straighten the knee, then the hip, and finally the trunk. The results suggest that the trunk might be positioned over the feet for stability when the extension part of the task was initiated, reminiscent of the stability-driven strategy described by Hughes et al. (p. 158)

It would seem natural to conclude this might also have a relationship with quadriceps strength but linear regression did not support this in our sample (Figure 12). It seems participants were able to create compensatory biomechanical adjustments which were unrelated to their maximal isometric knee extensor strength (Figure 2). Dynamic expression of strength in the form of efficient motor recruitment (magnitude, speed, and neural organization of muscle firing) all could influence their performance but were not accounted for in the study’s measures.

Emphasis is placed on lack of lateral stability in fall-prone populations since it inherently contributes to mistranslated energy for forward propulsion and increased risk of lateral falls with potentially severe consequences. Researchers including Chen et al. [40] have proposed that reductions in forward velocity could manifest as increased M/L velocities but their studies have not irrefutably demonstrated that older, fall-prone adults have significantly greater magnitudes than controls. A study by Aberg et al. [36] concluded that, while their M/L velocity parameters were nonsignificant between a group with fear of falling and an age-matched control group, the fear of falling group displayed a disproportionately higher sideways velocity due to the significant reduction in forward velocity.

While this study cannot make any claims on the baseline measurements of the participants' M/L velocity or momentum parameters, it is important to note that no M/L variables were found to be significant between bed heights for any STW event – with one exception: M/L trunk momentum at stance-off was found to differ between conditions with RM ANOVA (p=0.042). However, post-hoc testing did not confirm this. When the data were subsetted to examine only individuals with a MFS score of 50 or greater, HB and LB were found to be significantly different (p=0.038) with a mean difference of 0.660 kg-m/s. Stance-off kinematics were frequently confounded by individuals who began turning to sit down in the chair opposite the bed as early as this second step, although this was less common in more impaired subjects; therefore, the clinical importance is unclear.

In light of Aberg's findings about velocity relationships between A/P and M/L axes, additional analysis was carried out to determine if any velocity ratios were significantly different between bed heights. RM ANOVA was conducted on M/L:A/P velocity ratios at swing-off events and M/L:A/P peak velocity ratios.[2] No significance was detected in either, respectively, with or without the removal of outliers (Figure 13 and Figure 14). These metrics were considered adequate to represent the most posturally-challenging aspects of STW in this study. While mobility measures appear to differ in some respects between bed heights, these results lend evidence that participants appear to exhibit the same degree of control among all conditions. It should be noted that negative ratios indicate a posterior A/P velocity. All M/L kinematics were analyzed using absolute value since direction of sway did not matter. The more impaired study participants often used a whole-foot “stepping” or shuffling motion when initiating gait and did not create the typical “push” using the forefoot. These cases could create posterior motion at toe-off.

Lastly, if only the representative variables of the PCA were presented in the results and discussion (Table 3), similar conclusions could be made albeit with less detail: the only significant variable in the dimension reduction was a time variable (component 1 of the PCA). An A/P velocity at the first gait event was borderline significant as well (component 2 of the PCA). No other representative variables of the five other components were significant – a result that mostly agrees with the rest of the ANOVA findings in this study. In other words, temporal parameters matter most when considering bed height in a fall-prone population

[2] Peak signals were pulled from the entire signal outputted during the time period from peak anteroposterior ground reaction force through stance-off events.

with velocity parameters a close second. Momentum parameters have less influence on stability changes between differing bed heights. These conclusions represent 71% of the variance in the data.

4.3 Limitations

By allowing an unconstrained approach to STW, a large degree of biological variability was considered for analysis at the expense of controlling for confounders. In particular, this is evident in the lack of data related to side rail usage since it was not mandated.

Data were not collected on a comparable control group.

Difficulties in placing markers on overweight and obese subjects as well as the presence of safety vest used in the fall arrest system made it difficult to avoid artifact in marker placement.

This study did not seek to actually induce falls and, as such, it is not known if the experimental design was optimal to produce kinetic and kinematic values which could predict actual fall risks.

The data set as a whole contained more than 1000 individual motion trials. All of them required human-powered error checking and adjustment. A team of 5-6 individuals performed this over a one-year time period. Individual judgment variations in what constituted exact events and human fallibility in catching all errors must factor into the results obtained here. Additionally, many perfectly good files and data points had to be excluded due to a process that demanded any obvious discrepancies in analog data or motion data outputs be passed over.

This was due to the excessive time required to fix and reintegrate such data in light of the end mission.

Due to the recruitment site (Veteran’s Administration), the sample was skewed toward Caucasian men. Women made up only 31% of the sample and racial breakdown was 90% Caucasian.

4.4 Future Work

There are many options and directions for future work. Analyzing the overall variability of the A/P and M/L velocity and momentum signals during STW would provide more insight into kinematic aberrations associated with falls. Exploration of kinetic energy and/or velocity and/or momentum distributions during rising and gait initiation in all three components (and as they relate to temporal parameters) could reveal more detail of the interaction between stability and mobility from various seat heights, especially during the act of rising. Examining angular joint accelerations and peak angles at key events could provide additional insight into energy generation and use.

Studying extended ambulation following STW at various seat heights could more adequately assess published data findings of increased M/L velocities during established gait. Quantifying center of gravity in relation to center of pressure as well as the braking impulse following rising could provide additional avenues of stability measures.

Additional analysis of side rail trials subsetted to characterize kinematic differences between use and nonuse at a controlled bed height may also lend insight to theories regarding their presence as aids or hazards in actual hospitals.

Measuring hip extensor strength and using EMG on lower limb musculature could provide a deeper view into the involved physiology.

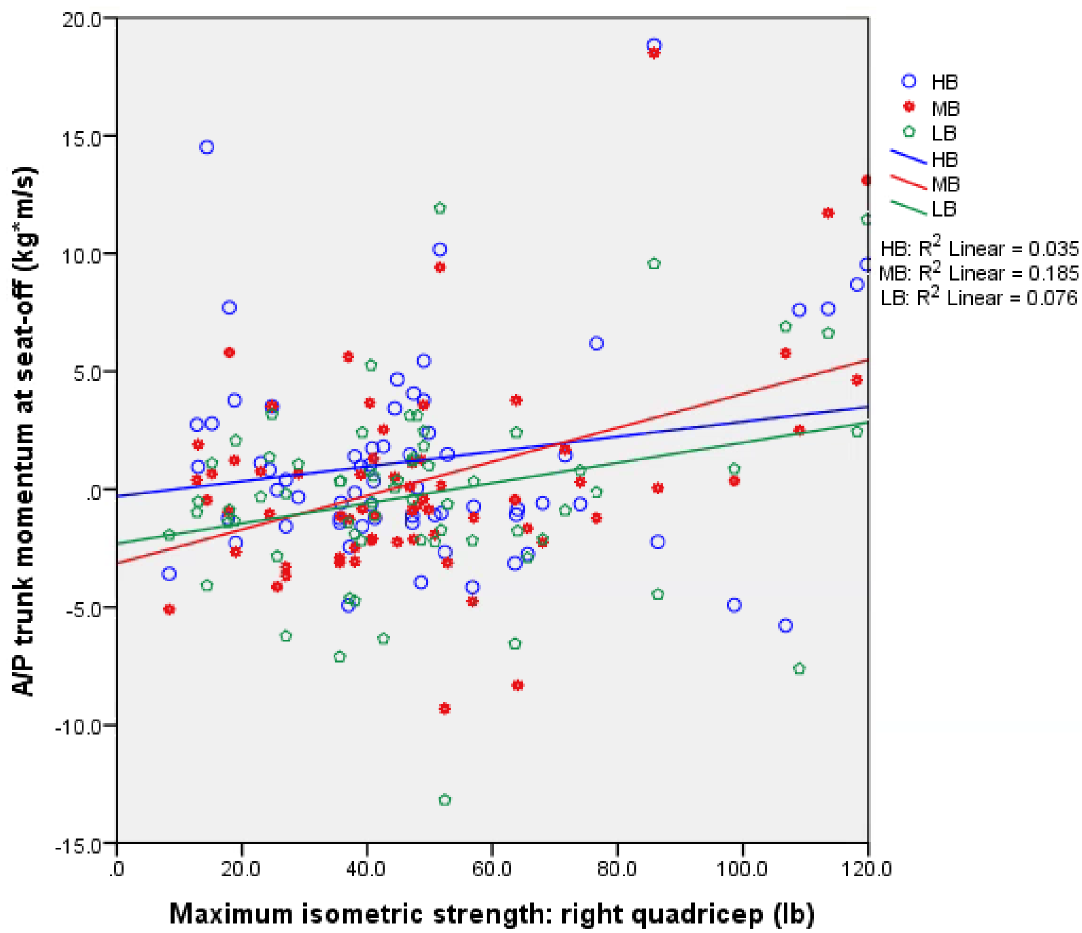


Figure 12. Scatter plot results of participant strength measure against momentum generation at seat-off.

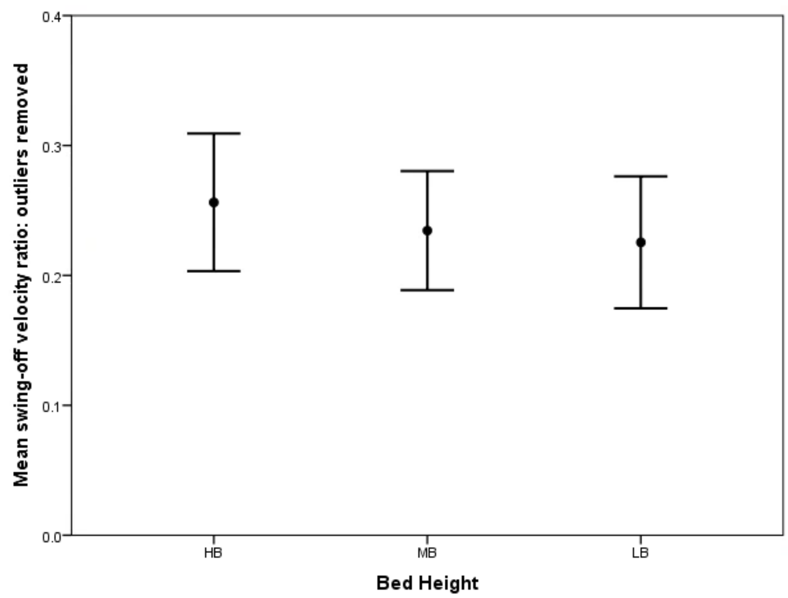


Figure 13. 95% CI for M/L:A/P velocity ratio at swing-off (outliers removed).

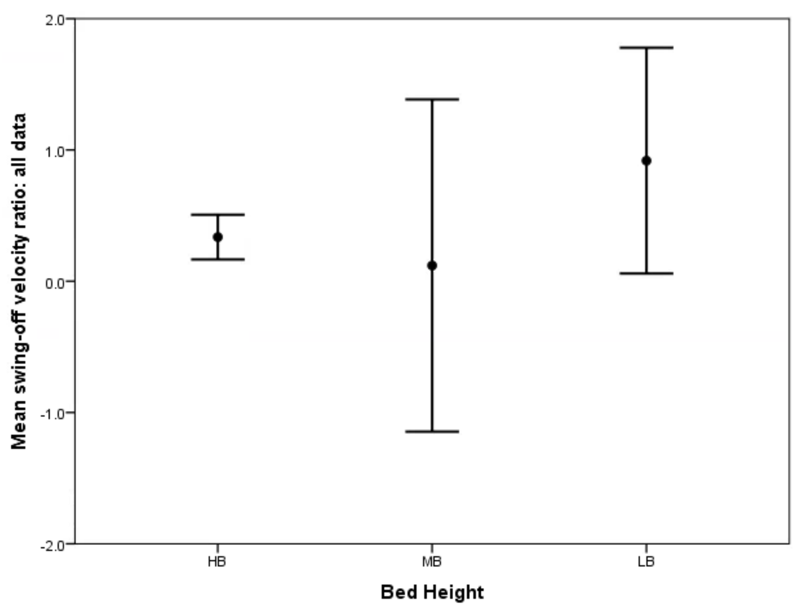


Figure 14. 95% CI for M/L:A/P velocity ratio at swing-off (outliers included).

CHAPTER 5

CONCLUSION

Bed height appears to play a role in kinematic and temporal fall risk parameters. The low bed height created the largest delays in temporal parameters after seat-off occurred but displayed no significant differences in A/P velocities or momentums at most key events. This indicates that low beds pose a more posturally challenging transition and require more time to accommodate balance impairments while using the same momentum strategy as the other bed heights. These results are illuminating in that a slightly lower bed height may confer significant mobility impairments when one considers the fact that the low bed was an average of only 2.5 (±0.1) inches below the medium bed and only 0.9 (±0.1) inches less than tibial plateau height.

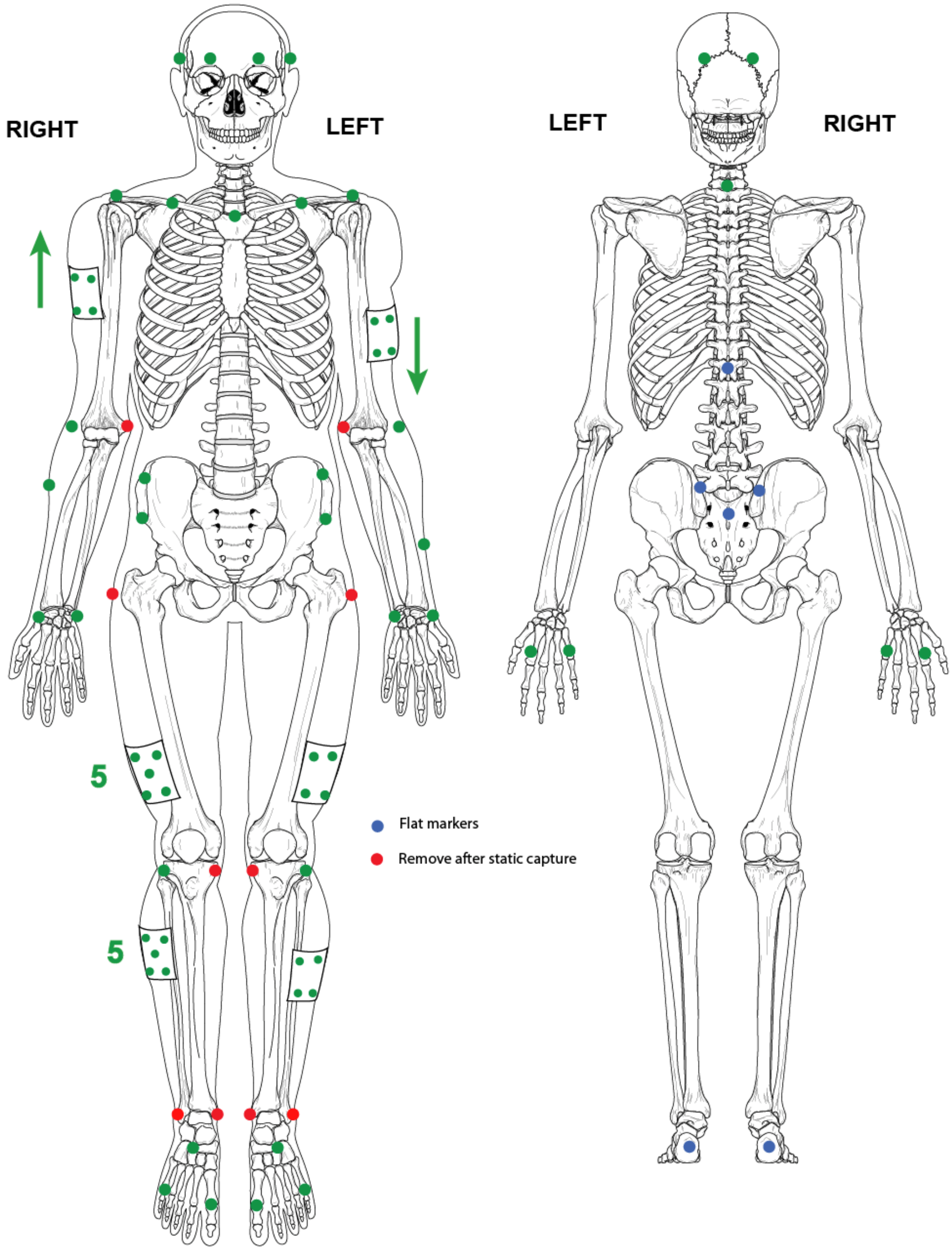
The medium bed height had the least significantly different kinematic or temporal parameter outcomes in pairwise comparisons with both other bed heights, indicating it may provide a middle ground for patients to utilize either stabilization or mobilization strategies. These two kinematic expressions are often seen as noncomplementary yet it is imperative to use both in a fluid manner for successful STW [38]. Healthy people do this naturally but this ability decreases with age and disease. The medium bed height allowed for time-to-rise phases to be longer or shorter as needed yet gait initiation to be faster. It also provided the greatest increase in velocity between seat-off and swing-off. Thus the potential beauty of a medium bed height is that it allowed participants to rise at a rate which felt stable yet enter into gait with a fairly high degree of fluidity.

The high bed height had many of the same kinematic and temporal characteristics as the medium bed height with slightly faster event times. Since mediolateral kinematic parameters were statistically the same as medium bed heights, high bed heights in the range of 3 inches beyond tibial plateau height may pose little hazard for individuals with fall risk during bed egress despite the fact that many had to scoot forward on the bed and/or “touch down” to the floor upon exit.

This study supports qualitative literature findings that low bed heights do not appear to reduce fall risks during bed exit; rather, they may exacerbate them. Lastly, since temporal parameters showed the greatest significance between bed heights, and regression analysis correlated the TUG test as most predictive in these parameters, the simple TUG metric gleaned from a patient’s medical history may assist clinical staff in assessing an ambulatory patient’s need for a medium to slightly high bed height setting. This type of application takes virtually no time to perform and may yield some improvement in the puzzle of fall prevention.

APPENDIX A

MARKER PLACEMENT


RIGHT
LEFT
LEFT
RIGHT
5
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Flat markers
Remove after static capture

APPENDIX B

MORSE FALL SCALE QUESTIONNAIRE

Morse Fall Scale

(Adapted with permission, SAGE Publications)

The Morse Fall Scale (MFS) is a rapid and simple method of assessing a patient's likelihood of falling. A large majority of nurses (82.9%) rate the scale as "quick and easy to use," and 54% estimated that it took less than 3 minutes to rate a patient. It consists of six variables that are quick and easy to score, and it has been shown to have predictive validity and interrater reliability. The MFS is used widely in acute care settings, both in the hospital and long term care inpatient settings.

Item	***Scale***	***Scoring***
1. History of falling; immediate or within 3 months	No 0 Yes 25	______
2. Secondary diagnosis	No 0 Yes 15	______
3. Ambulatory aid Bed rest/nurse assist Crutches/cane/walker Furniture	0 15 30	______
4. IV/Heparin Lock	No 0 Yes 20	______
5. Gait/Transferring Normal/bedrest/immobile Weak Impaired	0 10 20	______
6. Mental status Oriented to own ability Forgets limitations	0 15	______

The items in the scale are scored as follows:

History of falling: This is scored as 25 if the patient has fallen during the present hospital admission or if there was an immediate history of physiological falls, such as from seizures or an impaired gait prior to admission. If the patient has not fallen, this is scored 0. Note: If a patient falls for the first time, then his or her score immediately increases by 25.

Secondary diagnosis: This is scored as 15 if more than one medical diagnosis is listed on the patient's chart; if not, score 0.

Ambulatory aids: This is scored as 0 if the patient walks without a walking aid (even if assisted by a nurse), uses a wheelchair, or is on a bed rest and does not get out of bed at all. If the patient uses crutches, a cane, or a walker, this item scores 15; if the patient ambulates clutching onto the furniture for support, score this item 30.

Intravenous therapy: This is scored as 20 if the patient has an intravenous apparatus or a heparin lock inserted; if not, score 0.

Gait: A *normal gait* is characterized by the patient walking with head erect, arms swinging freely at the side, and striding without hesitant. This gait scores 0. With a *weak gait* (score as 10), the patient is stooped but is able to lift the head while walking without losing balance. Steps are short and the patient may shuffle. With an impaired gait (score 20), the patient may have difficulty rising from the chair, attempting to get up by pushing on the arms of the chair/or by bouncing (i.e., by using several attempts to rise). The patient's head is down, and he or she watches the ground. Because the patient's balance is poor, the patient grasps onto the furniture, a support person, or a walking aid for support and cannot walk without this assistance.

Mental status: When using this Scale, mental status is measured by checking the patient's own self assessment of his or her own ability to ambulate. Ask the patient, "Are you able to go the bathroom alone or do you need assistance?" If the patient's reply judging his or her own ability is consistent with the ambulatory order on the Kardex®, the patient is rated as "normal" and scored 0. If the patient's response is not consistent with the nursing orders or if the patient's response is unrealistic, then the patient is considered to overestimate his or her own abilities and to be forgetful of limitations and scored as 15.

Scoring and Risk Level: The score is then tallied and recorded on the patient's chart. Risk level and recommended actions (e.g. no interventions needed, standard fall prevention interventions, high risk prevention interventions) are then identified.

Important Note: The Morse Fall Scale should be calibrated for each particular healthcare setting or unit so that fall prevention strategies are targeted to those most at risk. In other words, risk cut off scores may be different depending on if you are using it in an acute care hospital, nursing home or rehabilitation facility. In addition, scales may be set differently between particular units within a given facility.

Sample Risk Level

Risk Level	MFS Score	Action
No Risk	0 - 24	Good Basic Nursing Care
Low Risk	25 - 50	Implement Standard Fall Prevention Interventions
High Risk	≥ 51	Implement High Risk Fall Prevention Interventions

ACRONYMS AND DATA DICTIONARY

Keyword	Description
A/P	Anteroposterior
ANOVA	Analysis of variance
BoS	Base of support
CoG	Center of gravity
CoM	Center of mass
DV	Dependent variable
Extension phase	The phase of STW which occurs between seat-off and first toe-off
GI	Gait initiation
GRF	Ground reaction force
HB	High bed
Heel-off	The point in time when the foot initiating gait begins the lateral weight transferrance to the opposite side via lifting the heel
IV	Independent variable
LB	Low bed
LLL	Lower leg length
M/L	Mediolateral
MB	Medium bed
MFS	Morse Fall Scale
N	The fixed reference frame (global)
P	The reference frame associated with the pelvis segment
PCA	Principal Component Analysis
RM	Repeated measures
S	The reference frame associated with a body segment (general)
Stance-off	The second occurring toe-off following the rising phase in STW; the end event of the task
STS	Sit-to-stand
STW	Sit-to-walk
Swing-off	The first occurring toe-off following the rising phase in STW
Time ratio	The ratio created between the extension phase duration as a percentage of the total task time
Total task time	The duration of the entire STW task: defined as starting at movement initiation and ending at stance-off
TUG	Timed up and go
V3D	Visual3D

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